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Method of creating images in computer tomography, and
CT device for carrying out the method

5   The invention relates to a method of creating images in
computer tomography, in which an object under
examination is scanned with the aid of a beam
originating from a focus and a flat detector array, the
output data determined is filtered in a suitable way
and backprojected in order to obtain at least one slice
10   which represents the absorption values of the section
of the object under examination. Furthermore, the
invention relates to a CT device which is suitable for
carrying out this method.

15   Methods of this type have become known under the term
"filtered backprojection" but, as 3D methods, that is
to say in conjunction with a matrix-like detector
array, do not furnish an image quality which is judged
to be adequate in practice, since what are known as
20   "cone beam artefacts" occur because of the conical x-
ray beam.

In addition, the disadvantage with this method is that
redundant data, such as are produced during spiral
25   scanning with low table advance as a result of multiple
irradiation of one and the same voxel, is not used,
which results in the radiation dose administered to the
object under examination being used only incompletely
for imaging.
30
Furthermore, there are thoughts relating to proceeding,
in conjunction with 2D methods for image
reconstruction, in such a way that preliminary images
are calculated in large numbers by means of "filtered
35   backprojection" from initial data which originates from
sections of the focal path which are intrinsically
inadequate for image reconstruction, the preliminary
images being reformatted to form a final slice only in

a second step. These 2D methods are less useful for detector arrays with a large width, that is to say a great extent in the direction of the system axis, since an extremely large number of preliminary images then 5   has to be processed, which is a problem even when high computing power is available.

In all the aforementioned methods, the problem arises that, because of possible data redundancy which occurs 10   during the scanning of the object under examination, image artefacts are produced which have a detrimental influence on the image quality.

The invention is based on the object of specifying a 15   method of the type mentioned in the introduction which permits the image quality to be increased.

According to the invention, this object is achieved by the features of the independent patent claims. Special 20   configurations and improvements emerge from the features of the subclaims.

The inventors have discovered that in image processing it is firstly necessary to take account of the 25   redundancies which occur with respect to a respectively considered voxel in the region under examination, in particular during the backprojection of the filtered data. Secondly, however, it is also necessary to take into account how the ray is positioned in the beam. For 30   example, it has a decisively positive effect on the image quality if rays which are located centrally, based on the extent of the beam in the direction of the axis of rotation, are taken more into account when creating the image than rays which are positioned only 35   marginally in relation to the beam, based on the same extent.

Accordingly, the inventors propose an improved method for creating images in computer tomography which has the following method steps:

- to scan an object under examination with a beam
5 originating from at least one focus and with a flat detector array having a large number of distributed detector elements for detecting the rays of the beam, the at least one focus is rotated relative to the object under examination on at least one focal
10 path running around the object under examination with opposite detector array, the detector elements of the detector array supplying initial data which represents the attenuation of the rays as they pass through the object under examination,
15 - the initial data is filtered,
- the filtered initial data is then backprojected three-dimensionally in order to produce at least one slice of a layer of the object under examination which has a layer thickness, the slice representing
20 absorption values, obtained from the initial data, of the voxels belonging to the layer of the object under examination for the radiation of the beam, and
- during the backprojection, the rays are weighted as a function of their position in the beam.

25

In a particular embodiment, the filtering is carried out in the direction of the tangent to the focal path belonging to the respective focal position. It has been shown that, with this filtering direction, a
30 particularly high image quality can be achieved. The selection of this filtering direction is based on the finding that the abovedescribed 2D method permitting a high image quality and based on preliminary images would, so to speak, change into a 3D method if the
35 sections of the focal path on which the calculation of preliminary images is based were to be shortened so severely that they would cover only a single projection whose data would then be filtered in the direction of

the tangent to the focal path, and in that it would then be possible to expect that such a 3D method would then permit a comparatively good image quality as the 2D method.

If the beam has an extent in the direction of rotation and an extent in the direction of the axis of rotation, then it is advantageous to weight rays which are arranged centrally in the beam, as based on the extent of the beam in the direction of the axis of rotation, to a greater extent than the rays which are arranged close to the edge in the beam, as based on the extent of the beam in the direction of the axis of rotation.

The method according to the invention may then be carried out particularly easily if, before filtering, a conversion of the initial data obtained in fan beam geometry in the form of rays $P(\alpha,\beta,q)$ into parallel data present in parallel beam geometry in the form of rays $P(\theta,\beta,q)$ (azimuthal "rebinning") or $P(\theta,p,q)$ (complete "rebinning", that is to say azimuthal and radial "rebinning") is performed. In this case, with reference to fig. 3

α	is the focal angle
β	is the fan angle
q	is the row index of the detector system corresponding to the z coordinate,
$\theta = \alpha + \beta$	is the parallel fan angle,
$p = R_F sin(\beta)$	is the parallel coordinate corresponding to the distance of the ray from the axis of rotation (system axis), and
R_F	is the radius of the focal path.

According to a preferred embodiment of the invention, the backprojection of the parallel data is carried out in that, in the course of the backprojection for each voxel $V(x,y,z)$, for each $\theta \in [o,\pi]$ for the rays

$P(\theta+k\pi,\widetilde{\beta},q)$ and $P(\theta+k\pi,\widetilde{p},q)$ whose projection along the system axis goes through (x,y), the sum

$$P_{x,y,z}(\theta) = \sum_{k}\sum_{q} W \cdot h\left(d_{x,y,z}\left(\theta+k\pi,\left\{\begin{matrix}\widetilde{p}\\\widetilde{\beta}\end{matrix}\right\},q\right)\right) \cdot P\left(\theta+k\pi,\left\{\begin{matrix}\widetilde{p}\\\widetilde{\beta}\end{matrix}\right\},q\right)$$

is formed, where

x,y,z are the coordinates of the respective voxel $V(x,y,z)$,

k is a whole number corresponding to the number of half revolutions of the focus included in the reconstruction,

$\widetilde{p}$ are the parallel coordinates of those rays whose projections along the system axis run through the coordinates (x,y) of the respective voxel $V(x,y,z)$,

$\widetilde{\beta}$ are the fan angles of those rays whose projections along the system axis run through the coordinates (x,y) of the respective voxel $V(x,y,z)$,

h is a weighting function determining the layer thickness of the layer of the object under examination represented in the slice produced,

d is a function which is equal to the distance of the respective ray from the corresponding voxel $V(x,y)$ and or is dependent on the distance of the respective ray from the corresponding voxel $V(x,y)$, and

W represents a weighting function which weights rays with a large parallel fan angle θ less than rays with a small parallel fan angle θ.

In this case, the notation $\left\{\begin{matrix}\widetilde{p}\\\widetilde{\beta}\end{matrix}\right\}$ expresses the fact that the formation of the sum can optionally be carried out for rays obtained by means of azimuthal "rebinning" or complete "rebinning", the filtering tangential to the

focal path in the case of azimuthal "rebinning" being filtering in the β direction and, in the case of complete "rebinning", being filtering in the p direction.

5

Because of the summing both over k and over q, it is then ensured that all the rays running through one and the same voxel are taken into account and the radiation dose supplied to the object under examination is thus

10   used completely.

A particularly preferred embodiment provides that, for the purpose of the backprojection of the parallel data, the sum

15

$$H = \sum_{k}\sum_{q} W \cdot h \cdot \left(d_{x,y,z}\left(\theta + k\pi, \left\{ \frac{\tilde{p}}{\tilde{\beta}} \right\}, q \right) \right)$$

normalized to the sum H of the weights h

$$P_{x,y,z}(\theta) = \frac{1}{H} \sum_{k}\sum_{q} W \cdot h \left(d_{x,y,z}\left(\theta + k\pi, \left\{ \frac{\tilde{p}}{\tilde{\beta}} \right\}, q \right) \right) \cdot P\left(\theta + k\pi, \left\{ \frac{\tilde{p}}{\tilde{\beta}} \right\}, q \right)$$

20

is formed. This procedure permits an image quality which is improved again, since possible overemphasis of voxels which are struck by more rays than other voxels

25   is eliminated, and thus corresponding artefacts are avoided. The CT value of the respective voxel is obtained by summing over θ.

According to the invention, the weighting function can

30   be a function of the parallel fan angle with $W(\theta+k\pi)$, and preferably represent a smooth function which has the value 1 for central rays, as based on the extent of the fan of rays in the direction of the axis of rotation or z direction, and tends to 0 for marginal

35   rays.

If the detector array has detector elements arranged in the manner of rows, then the weighting can be represented by a function of the row number $W(q)$, a smooth function also preferably being used here which, for rays with a centrally placed detector row or a plurality of centrally placed detector rows, has a high value, preferably the value 1, and for rays arranged at the margin tends to 0; for example this can be a $\cos^2$ function.

The method according to the invention can be used, for example when, according to one variant of the invention, the focal path is a circular path (tomogram scanning). According to a preferred variant of the invention, however, the focal path is a spiral path, which is brought about by the focus being moved on a circular path about the system axis and, at the same time, there being a relative movement between focus and object under examination in the direction of the system axis. On the basis of such spiral scanning, even relatively large volumes of the object under examination can be examined without problems.

In the case of tomogram scanning, k is normally $k=1$ or $k=2$; in the case of spiral scanning, k is selected while taking account of the relative displacement in the direction of the system axis carried out per full revolution, such that the region to be imaged of the object under examination is registered completely.

According to the basic idea of the invention, the inventors also propose to improve a CT device for scanning an object under examination with a beam originating from at least one focus and with a flat detector array having a large number of distributed detector elements for detecting the rays of the beam, the at least one focus moving relative to the object

under examination on at least one focal path running
around the object under examination with the opposite
detector array, to the effect that at least means for
collecting detector data, filtering and backprojection
5   are provided, which carry out the method described
above, these means preferably and as far as possible
being implemented by programs or program modules.

The invention will be explained in more detail below
10 using an exemplary embodiment which is illustrated in
the appended schematic drawings, in which:

Fig. 1   shows a CT device having a plurality of rows of
         detector elements in a partly perspective,
15   partly block-diagram illustration;
Fig. 2   shows a longitudinal section through the device
         according to Fig. 1;
Fig. 3   shows a diagram illustrating "rebinning";
Fig. 4   shows scanning with opposite focus and detector
20   in the beam with a rotating gantry;
Fig. 5   shows the course of two different weighting
         functions W;
Fig. 6   shows scanning with opposite focus and detector
         in the beam with the gantry rotating and moving
25   in the z direction.

Figs 1 and 2 illustrate a CT device of the third
generation suitable for carrying out the method
according to the invention. Its measurement
30 arrangement, designated overall by 1, has an x-ray
source, designated overall by 2, with a beam aperture
stop 3 arranged in front of the latter, close to the
source (Fig. 2) and a detector system 5 formed as a
two-dimensional array of a plurality of rows and
35 columns of detector elements - one of these is
designated 4 in Fig. 1 - with a beam aperture stop 6 in
front of the latter, close to the detector (Fig. 2). In
Fig. 1, for clarity, only eight rows of detector

elements 4 are shown. However, as indicated by dots in Fig. 2, the detector system 5 has further rows of detector elements 4.

5 The x-ray source 2 with the aperture stop 3, on the one hand, and the detector system 5 with the aperture stop 6, on the other hand, are fitted, in the manner which can be seen from Fig. 2, to a rotary frame 7, opposite each other in such a way that a pyramidal x-ray beam

10 originating from the x-ray source 2 during operation of the CT device and masked by the adjustable aperture stop 3, whose marginal rays are designated 8, strikes the detector system 5. In this case, the aperture stop 6 is adjusted in accordance with the cross section of

15 the x-ray beam set by the aperture stop 3 such that only that region of the detector system 5 which can be struck directly by the x-ray beam is exposed. In the operating mode illustrated in Figs 1 and 2, these are eight rows of detector elements 4, which are referred

20 to as active rows in the following text. The further rows, indicated by dots, are covered by the aperture stop 6 and therefore not active.

Each row of detector elements 4 has a number K of

25 detector elements, $\beta_k = \beta_1$ to β_K being the channel index and each detector element being assigned a fan angle β_k. The fan angle of the central detector element is equal to 0; the fan angles of the two outermost detector elements are $\beta_1 = +\beta_{max}$ and $\beta_K = -\beta_{max}$.

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The active rows L_q of detector elements 4 are designated L_1 to L_Q, $q = 1$ to Q being the row index which, in the case of the exemplary embodiment described, corresponds so to speak to the z coordinate.

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The x-ray beam has the cone angle φ plotted in Figs 1 and 2, which is the opening angle of the x-ray beam in a plane containing the system axis Z and the focus F.

The opening angle of the x-ray beam in a plane at right angles to the system axis Z and containing the focus F (fan opening angle) is $2\beta_{max}$ and is plotted in Fig. 1.

5   The rotary frame 7 can be set rotating by means of a drive device 22 about a system axis designated by Z. The system axis Z runs parallel to the z axis of a three-dimensional rectangular coordinate system illustrated in Fig. 1.

10

The columns of the detector system 5 likewise run in the direction of the z axis, while the rows, whose width b is measured in the direction of the z axis and is 1 mm, for example, run transversely with respect to
15   the system axis Z and the z axis.

In order to be able to move an object under examination, for example a patient, into the beam path of the x-ray beam, a mounting device 9 is provided,
20   which can be displaced parallel to the system axis Z, that is to say in the direction of the z axis, to be specific in such a way that there is synchronization between the rotational movement of the rotary frame 7 and the translational movement of the mounting device,
25   with the effect that the ratio between translation and rotational speed is constant, this ratio being adjustable by a desired value for the advance v of the mounting device per revolution of the rotary frame being selected.

30

It is therefore possible for a volume of an object under examination located on the mounting device 9 to be examined in the course of a volume scan, it being possible for the volume scan to be carried out in the
35   form of spiral scanning, with the effect that, with simultaneous rotation of the measuring unit 1 and translation of the mounting device 9, a large number of projections from different projection directions is

recorded by means of the measuring unit for each revolution of the measuring unit 1. During the spiral scanning, the focus F of the x-ray source moves relative to the mounting device 9 on a spiral path

5 designated S in Fig. 1. The spiral scanning must extend in the α direction over at least $\pi+2\beta_{max}$, in order to permit the complete reconstruction of a CT image for each row of detector elements, but can also be longer as desired within the technical limits of the CT

10 device.

However, because of the fact that a plurality of rows of detector elements 4 are present, a volume of the object under examination can also be examined in the

15 course of what is known as a tomogram scan, in which no relative movement takes place in the direction of the z axis between the measuring unit 1 and mounting device 9 (v = 0). In the case of the tomogram scan, the size of the volume examined is therefore determined by the

20 number of active rows of detector elements 4. During a tomogram scan, the focus F moves on a circular focal path which lies in a plane designated the mid-plane below.

25 The tomogram scanning can be carried out in the form of a part revolution or in the form of a complete revolution, the part revolution covering a part revolution interval of at least $\pi+2\beta_{max}$ (half a revolution plus the fan opening angle), which permits

30 complete reconstruction of a CT image, while a full revolution covers 2π.

The measured data read out in parallel from the detector elements of each active row of the detector

35 system 5 during the spiral or tomogram scanning and corresponding to the individual projections $P(\alpha, \beta, q)$ in fan beam geometry is subjected to digital/analog

conversion in a data conditioning unit 10, serialized and transmitted to an image computer 11.

5　 Following preconditioning of the measured data in a preconditioning unit 12 belonging to the image computer 11, the resultant data stream passes to a slice reconstruction unit 13, which uses the measured data to reconstruct slices of desired layers of the object under examination in accordance with a method according 10　 to the invention based on "filtered backprojection" and yet to be described in detail.

The CT images are composed of pixels (pixel = picture element) assembled in the manner of a matrix, the 15　 pixels being assigned to the respective image plane, each pixel being assigned a CT number in Hounsfield units (HU) and the individual pixels being displayed in accordance with a CT number/gray value scale in a gray value corresponding to their respective CT number. In 20　 this case, each pixel illustrates a voxel (voxel = volume element) of the layer illustrated in the CT image and belonging to the object under examination. Since, because of the multiplicity of rows of the detector system 5 and, if appropriate, the spiral 25　 scanning, measured data with respect to a plurality of layers of the object under examination is obtained, 3D data is available, which is subjected to 3D backprojection within the context of the invention. As the final result, 3D image data is available in the 30　 form of a three-dimensional matrix, for example with the axes x, y, z, each element in the matrix corresponding to a voxel $V(x, y, z)$ and containing the gray value corresponding to the associated CT number. Those elements of the three-dimensional matrix which 35　 have the same x, y, or z value then in each case represent a planar slice of the layer of the object under examination corresponding to the defining x, y, or z value.

The images reconstructed by the slice reconstruction unit 13 are displayed on a display unit 16, for example a monitor, connected to the image computer 11.

5

The x-ray source 2, for example an x-ray tube, is supplied with the necessary voltages and currents, for example the tube voltage U, by a generator unit 17. In order to be able to set these to the respectively

10   necessary values, the generator unit 17 is assigned a control unit 18 with keyboard 19, which permits the necessary adjustments.

In addition, the other operation and control of the CT

15   device is carried out by means of the control unit 18 and keyboard 19, which is illustrated by the fact that the control unit 18 is connected to the image computer 11.

20   Inter alia, the number Q of active rows of detector elements 4, and therefore the position of the aperture stops 3 and 6, can be adjusted, for which purpose the control unit 18 is connected to adjusting units 20 and 21 assigned to the aperture stops 3 and 6.

25   Furthermore, the rotation time τ needed by the rotary frame 7 for a complete revolution can be adjusted, which is illustrated by the fact that the drive unit 22 assigned to the rotary frame 7 is connected to the control unit 18.

30

Although it is possible in principle to implement the method according to the invention in fan beam geometry as well, the CT device described is preferably operated in a mode in which the method according to the

35   invention is implemented in parallel beam geometry.

Accordingly, the data obtained during the scanning of the region of the body of the patient which is relevant

to the respective examination by means of spiral or
tomogram scanning in fan beam geometry is first of all
converted, in a manner known per se, into data in
parallel beam geometry by means of a method generally
5   designated "rebinning". This conversion is based on
resorting the data obtained in fan beam geometry in
such a way that rays are removed from different
projections recorded in fan beam geometry, and joined
together to form a projection in parallel beam
10   geometry. In parallel beam geometry, data from an
interval with a length π is sufficient to be able to
reconstruct a complete image. In order to obtain this
data, data in fan beam geometry from an interval of
length $\pi+2\beta_{max}$ must nevertheless be available.
15

A projection in parallel beam geometry is illustrated
in Fig. 3. Accordingly, all n parallel rays RP_1 to RP_N
of this projection assume the parallel fan angle θ with
respect to the x axis of the coordinate system which is
20   illustrated in Fig. 3 and coincides with that according
to Fig. 1.

By using the parallel ray RP_1 illustrated by a
continuous line in Fig. 3, the transition from fan beam
25   to parallel been geometry will be explained below.

The parallel ray RP_1 originates from a projection
obtained in fan beam geometry for the focal position F_1
located on the focal path S. The central ray RF_{z1}
30   belonging to this projection in fan beam geometry,
running through the axis of rotation 14 and therefore
the z axis of the coordinate system, is likewise
plotted in Fig. 3. The focal position F_1 corresponds to
the focal angle α_1; this is the angle formed by the x
35   axis and the central ray RF_{z1}. As compared with the
central ray RF_{z1}, the ray RP_1 has the fan angle β. It is
thus easy to see that, for the parallel fan angle θ it
is true that $\theta = \alpha + \beta$.

The beam spacing p from the axis of rotation 14 or the z axis, measured at right angles to the respective parallel ray, is given by $p=R_F \sin(\beta)$.

5

As becomes clear by using the central ray RP_Z illustrated in a thickened line in Fig. 3 and extending through the axis of rotation 14 and the x axis, this ray is the central ray of a projection in fan beam geometry recorded in fan beam geometry for the focal position F_Z at the focal angle α_Z. Since it is true that $\beta=0$ for the central ray of a projection recorded in fan beam geometry, it becomes clear that the following is true of the case of central rays:

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Depending on whether an azimuthal or complete "rebinning" is carried out, the parallel projections are present in the form
$$P(\alpha, \beta, q)$$
or in the form
$$P(\theta, p, q)$$
where

α	is the focal angle
β	is the fan angle
q	is the row index of the detection system corresponding to the z coordinate,
$\theta=\alpha+\beta$	is the parallel fan angle
$p=R_F \sin(\beta)$	is the parallel coordinate corresponding to the ray spacing from the axis of rotation (system axis), and
R_F	is the radius of the focal path.

In an operating mode corresponding to a first embodiment of the method according to the invention, which can be selected by means of the keyboard 19, the CT device described operates on the basis of projections obtained by azimuthal "rebinning". The data corresponding to these projections is filtered in the β

direction, that is to say in each case in the direction of the tangent T belonging to the focal position of the central ray of the respective parallel projection (see fig. 3), specifically by using one of the filter kernels customary in computer tomography, e.g. a Shepp-Logan or Ramachandran-Lakshminarayanan kernel.

The parallel data filtered in this way is then backprojected in that, in the course of the backprojection, for each voxel $V(x,y,z)$ for each $\theta \in [0,\pi]$ for the rays $P(\theta + k\pi, \tilde{\beta}, q)$ whose projection along the system axis goes through (x,y), the sum

$$P_{x,y,z}(\theta) = \sum_k \sum_q W(q) \cdot h\big(d_{x,y,z}(\theta + k\pi, \tilde{\beta}, q)\big) \cdot P\big(\theta + k\pi, \tilde{\beta}, q\big)$$

is formed, where

x,y,z are the coordinates of the respective voxel $V(x,y,z)$,

k is a whole number corresponding to the number of half revolutions of the focus included in the reconstruction,

$\tilde{\beta}$ are the fan angle as of those rays whose projections along the system axis run through the coordinates (x,y) of the respective voxel $V(x,y,z)$, and

h is a weighting function which determines the layer thickness of the layer of the object under examination illustrated in the slice produced, and

d is a function which is equal to the distance of the respective ray from the corresponding voxel $V(x,y)$ or is dependent on the distance of the respective ray from the corresponding voxel $V(x,y)$, and

$W(q)$ represents a weighting function which weights rays which are positioned peripherally or at

least decentrally in the beam, as based on the
extent in the direction of the axis of
rotation, less than rays which are arranged
centrally.

5

As a result of the selected filtering direction and as
a result of the summation both over k and over q,
firstly cone beam artefacts are avoided, in the
interest of a high dose utilization, all the rays
10 running through a voxel $V(x,y,z)$ are taken into
account, but in addition, the positioning of the ray in
the beam is also taken into account by means of
appropriate weighting.

15 The absorption value $\mu_{x,y,z}$ associated with a voxel
$V(x,y,z)$ is obtained by means of summation over θ over
at least half a revolution, that is to say by forming

$$\mu_{x,y,z} = \sum_{\theta} P_{x,y,z}(\theta)$$

20

The CT number corresponding to the respective
absorption value is determined in a conventional way
from the absorption value.

25 In this case, different weighting functions h and
different functions d can be set by means of the
keyboard 19.

For example, a triangular or trapezoidal function is
30 suitable as a weighting function h.

The weighting function W is represented as a function
of the projection angle θ.

35 The function d can be the distance of the respective
. parallel ray from the voxel $V(x,y,z)$ or, instead, for
example the z (axial) component of this distance.

In a modification of the first operating mode described above, for the purpose of backprojection of the parallel data, the sum

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$$\mu_{x,y,z} = \sum_{\theta} P_{x,y,z}(\theta)$$

normalized to the sum H of the weights h

$$H = \sum_{k}\sum_{q} W(q) \cdot h\left(d_{x,y,z}\left(\theta + k\pi, \tilde{\beta}, q\right)\right)$$

10

is formed. This permits an image quality which is again improved, since the possible overemphasis of voxels which are "illuminated", that is struck by rays, in a
15 plurality of half revolutions is eliminated and thus corresponding artefacts are avoided. This redundancy occurs during spiral scans when the relative displacement taking place per full revolution of the measuring arrangement is so low (low pitch) that voxels
20 are irradiated repeatedly.

A second operating mode, which corresponds to a further embodiment of the method according to the invention and can be selected by means of the keyboard 19, differs
25 from the first operating mode in that the CT device described does not operate on the basis of projections obtained by means of azimuthal but by means of complete "rebinning". The data corresponding to these projections is filtered in the p direction in the case
30 of complete "rebinning", that is to say in each case likewise in the direction of the tangent T belonging to the focal position of the central ray of the respective parallel projection (see Fig. 3).

Accordingly, for the parallel data filtered in this way in the course of the backprojection, the sum

$$P_{x,y,z}(\theta) = \frac{1}{H}\sum_{k}\sum_{q}W(q)\cdot h\big(d_{x,y,z}(\theta + k\pi, \tilde{\beta}, q)\big)\cdot P(\theta + k\pi, \tilde{\beta}, q)$$

5

is formed, $\tilde{p}$ being the parallel coordinates of those rays whose projections along the system axis run through the coordinates *(x,y)* of the respective voxel *V(x,y,z)*.

10

In the case of the second operating mode, too, in a modification relating to the backprojection of the parallel data, a sum normalized to the sum *H* of the weights *h*

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$$P_{x,y,z}(\theta) = \sum_{k}\sum_{q}W(q)\cdot h\big(d_{x,y,z}(\theta + k\pi, \tilde{p}, q)\big)\cdot P(\theta + k\pi, \tilde{p}, q)$$

is formed, namely the sum

$$H = \sum_{k}\sum_{q}W(q)\cdot h\big(d_{x,y,z}(\theta + k\pi, \tilde{p}, q)\big)$$

20

In the case of the first and second operating modes described above, a function of the method according to the invention is provided in which, in conjunction with

25 a voxel *V(x,y,z)*, all the rays whose projection along the axis of rotation of 14 or the *z* axis goes through *x,y* are considered. Whether and to what extent these rays are taken into account is determined by the weighting functions *W* and *h* and the function *d*.

30

However, the CT device can also have further operating
modes which can be selected by means of the keyboard 19
and which correspond to those described previously,
with the difference that, for a given focal position,
5   the theoretical ray running through the respective
voxel $V(x,y,z)$ is determined and then, taking into
account the weighting function h and the function d in
the sum formation, only those rays which can actually
make a contribution to the sum are included in the
10 course of the backprojection.

In the exemplary embodiments described, the relative
movement between the measuring unit 1 and the mounting
device 9 is in each case produced by the mounting
device 9 being displaced. However, within the scope of
15 the invention, there is also the possibility of
arranging for the mounting position 9 to be fixed in
position and, instead, to displace the measuring unit
1. In addition, within the scope of the invention,
20 there is the possibility of producing the necessary
relative movement by displacing both the measuring unit
1 and the mounting device 9.

In Figure 4, the problem of weighting the rays in a 2D
25 representation is shown, simplified and illustrated
schematically. In this example, the gantry of the CT
with focus and detector in the 0° position is shown by
continuous lines and in the 180° position, rotated
about the z axis by 180°, is shown by dots. The
30 detector has 12 detector rows L1 to L12, which are each
struck by the associated rays S1 to S12 from the beam.
It should be pointed out that the spatial dimensions
shown are illustrated exaggerated with respect to the
extent of the detector, in order to be able to explain
35 the invention more clearly. The gantry shown moves on
a circular path here about the z axis or system axis.
Movement in the direction of the z axis does not take
place.

If the voxel V2 which lies centrally in the beam is
considered, this voxel will be penetrated by the
central rays S4 in both positions of the gantry.

5   According to the invention, the measurements from these
rays are highly weighted. At the same time, in the 0°
position, the voxel V1 which lies somewhat more
peripherally is penetrated in the 0° position by the
outer ray S1 and by the ray S4 lying further in. In

10   accordance with the weighting according to the
invention, the ray S1 will be given less weight than
the ray S4. If the likewise peripherally arranged voxel
V3 is considered, then the weighting here is reversed.
Since, here, the detector is constructed with detector

15   elements which are divided up into rows and columns in
the manner of a matrix, the weighting of the rays can
be coupled directly to the row number q.

In figure 5, located below, three examples of a course

20   according to the invention of weightings W_q are
illustrated. Common to all is the greater weighting of
the central rays with respect to the outer rays.
However, the functions are different. The dotted line
represents a course which drops off linearly on either

25   side of the centre, the dashed line results in
identical weighting over a large central region with a
tendency to drop off rapidly at the periphery, and the
continuous line shows a bell-like course which permits
a gentle transition between the extreme values 0 and 1.

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Figure 6 shows CT beams from a gantry of a spiral CT
respectively in the 0° position and adjacent 180°
position. In accordance with the essence of the spiral
CT, these positions are offset in relation to each

35   other in addition to the rotation in the z direction.
In this illustration, the voxel V1 is penetrated by the
ray S1 both in the 0° position and in the 180°
position. In accordance with this decentral position

of the ray S1 and the associated detector row L1 located on the outside in the detector, these two measurements are weighted lowly.   By contrast, the voxel V2 in the 0° position is penetrated by the ray S12 located decentrally, which is weighted lowly in this position for the voxel V2, but in the 180° position is penetrated by the central ray S6, which is weighted highly for this voxel V2. In accordance with the position of the rays S1 in the 0° and 180° positions, these rays are both weighted lowly in relation to the voxel V1. Overall, as a result of this additional weighting W illustrated here, the result is a substantial improvement in the image quality.

It should be pointed out that the conical x-ray beam has a rectangular cross section in the exemplary embodiment described here, but other cross-sectional geometries are also possible within the scope of the invention.

In connection with the exemplary embodiments described above, CT devices of the third generation are used, that is to say the x-ray source and the detector system are displaced jointly about the system axis during the image generation. However, the invention can also be used in connection with CT devices of the fourth generation, in which only the x-ray source is displaced about the system axis and interacts with a stationary detector ring, if the detector system is a multi-row array of detector elements.

The method according to the invention can also be used in CT devices of the fifth generation, that is to say CT devices in which the X-radiation originates from a plurality of foci of one or more x-ray sources displaced about the system axis, and not just from one focus, if the detector system has a multi-row array of detector elements.

The CT devices used in connection with the exemplary
embodiments described above have a detector system of
detector elements arranged in the manner of an
5   orthogonal matrix.   However, the invention can also be
used in connection with CT devices whose detector
system has a flat array with detector elements arranged
in another way.

10 The exemplary embodiments described above relate to the
medical application of the method according to the
invention.   However, the invention can also be applied
outside medicine, for example in checking luggage or in
material examination.

Patent claims

1.   A   method   of   creating   images   in   computer
      tomography, comprising the method steps:
1.1   to scan an object under examination with a beam
      originating from at least one focus and with a
      flat detector array having a large number of
      distributed detector elements for detecting the
      rays of the beam, the at least one focus is
      rotated relative to the object under examination
      on at least one focal path running around the
      object under examination with opposite detector
      array, the detector elements of the detector array
      supplying initial data which represents the
      attenuation of the rays as they pass through the
      object under examination,
1.2   the initial data is filtered,
1.3   the filtered initial data is then backprojected
      three-dimensionally in order to produce at least
      one slice of a layer of the object under
      examination which has a layer thickness, the slice
      representing absorption values, obtained from the
      initial data, of the voxels belonging to the layer
      of the object under examination for the radiation
      of the beam, and
1.4   during the backprojection, the rays are weighted
      as a function of their position in the beam.

2.   The method as claimed in claim 1, characterized in
      that the filtering is carried out in the direction
      of the tangent to the focal path belonging to the
      respective focal position.

3.   The method as claimed in either of claims 1 and 2,
      characterized in that the beam has an extent in
      the direction of rotation and an extent in the
      direction of the axis of rotation, rays which are

arranged centrally in the beam, as based on the extent of the beam in the direction of the axis of rotation, being weighted to a greater extent than the rays which are arranged close to the edge in the beam, as based on the extent of the beam in the direction of the axis of rotation.

4.   The method as claimed in one of claims 1 to 3, characterized in that, before filtering, a conversion of the initial data obtained in fan beam geometry in the form of rays $P(\alpha,\beta,q)$ into parallel data present in parallel beam geometry in the form of rays $P(\theta,\beta,q)$ or $P(\theta,p,q)$ is performed, where

α is the focal angle

β is the fan angle

q is the row index of the detector system corresponding to the z coordinate,

$\theta = \alpha + \beta$ is the parallel fan angle,

$p = R_F sin(\beta)$ is the parallel coordinate corresponding to the distance of the ray from the axis of rotation (system axis), and

R_F is the radius of the focal path.

5.   The method as claimed in claim 4, characterized in that the backprojection of the parallel data is carried out in that, in the course of the backprojection for each voxel $V(x,y,z)$, for each $\theta \in [o,\pi]$ for the rays $P(\theta + k\pi, \tilde{\beta}, q)$ and $P(\theta + k\pi, \tilde{p}, q)$ whose projection along the system axis goes through (x,y), the sum

$$P_{x,y,z}(\theta) = \sum_k \sum_q W \cdot h\left(d_{x,y,z}\left(\theta + k\pi, \left\{\frac{\tilde{p}}{\tilde{\beta}}\right\}, q\right)\right) \cdot P\left(\theta + k\pi, \left\{\frac{\tilde{p}}{\tilde{\beta}}\right\}, q\right)$$

is formed, where

x,y,z are the coordinates of the respective voxel
        V(x,y,z),

k   is a whole number corresponding to the
        number of half revolutions of the focus
        included in the reconstruction,

$\tilde{p}$ are the parallel coordinates of those rays
        whose projections along the system axis run
        through the coordinates (x,y) of the
        respective voxel V(x,y,z),

$\tilde{\beta}$ are the fan angles of those rays whose
        projections along the system axis run
        through the coordinates (x,y) of the
        respective voxel V(x,y,z),

h   is a weighting function determining the
        layer thickness of the layer of the object
        under examination represented in the slice
        produced,

d   is a function which is equal to the distance
        of the respective ray from the corresponding
        voxel V(x,y) or is dependent on the distance
        of the respective ray from the corresponding
        voxel V(x,y), and

W   represents a weighting function which
        weights rays with a large parallel fan angle
        θ less than rays with a small parallel fan
        angle θ.

6.   The method as claimed in claim 5, characterized in
      that, during the backprojection of the parallel
      data, the sum

$$H = \sum_k \sum_q W \cdot h \cdot \left(d_{x,y,z} \left(\theta + k\pi, \left\{ \frac{\tilde{p}}{\tilde{\beta}} \right\}, q \right) \right)$$

      normalized to the sum H of the weights h

$$P_{x,y,z}(\theta) = \frac{1}{H}\sum_{k}\sum_{q} W \cdot h\left(d_{x,y,z}\left(\theta + k\pi, \left\{\frac{\tilde{p}}{\tilde{\beta}}\right\}, q\right)\right) \cdot P\left(\theta + k\pi, \left\{\frac{\tilde{p}}{\tilde{\beta}}\right\}, q\right)$$

is formed.

7. The method as claimed in either of claims 5 and 6, characterized in that the weighting function represents a function of the parallel fan angle with $W(\theta+k\pi)$.

8. The method as claimed in claim 7, characterized in that the weighting function $W(\theta+k\pi)$ represents a smooth function which has the value 1 for centrally arranged rays in the beam, based on the parallel fan angle, and tends to 0 for rays arranged at the edge.

9. The method as claimed in either of claims 4 and 5, characterized in that the detector array has detector elements arranged in the manner of rows, and the weighting function represents a function of the row number $W(q)$.

10. The method as claimed in claim 8, characterized in that the weighting function $W(q)$ represents a smooth function which has the value 1 for rays to at least one centrally located detector row and tends to 0 for rays to detector rows at the edge.

11. The method as claimed in one of claims 1 to 9, characterized in that the focal path is a circular path.

12. The method as claimed in one of claims 1 to 9, characterized in that the focal path is a spiral path which is brought about by the focus being moved about the system axis on a circular path

and, at the same time, a relative movement between focus and object under examination in the direction of the system axis taking place.

5   13.   The method as claimed in one of claims 1 to 12, characterized in that the detector elements on the detector array are arranged distributed in rows and lines, preferably in the manner of a matrix.

10   14.   A CT device for scanning an object under examination, with a beam originating from at least one focus and with a flat detector array having a large number of distributed detector elements for detecting the rays of the beam, the at least one
15   focus moving relative to the object under examination on at least one focal path running around the object under examination with the opposite detector array, at least means for collecting detector data, filtering and
20   backprojection being provided as claimed in one of claims 1 to 13.

     15   The CT device as claimed in claim 14, characterized in that the aforementioned
25   functional means are at least partly implemented by programs or program modules.

Abstract

Method of creating images in computer tomography, and CT device for carrying out the method

The invention relates to a method for image reconstruction for computer tomography with a flat, extended detector, the rays being weighted during the backprojection as a function of their position in the beam.

In addition, the invention relates to a CT device for carrying out the above-mentioned method.
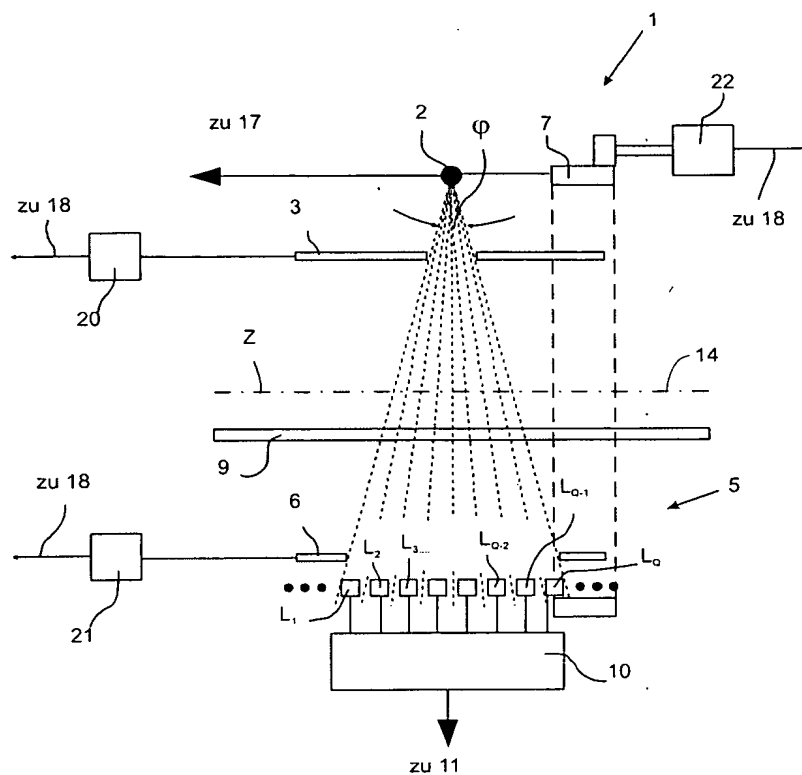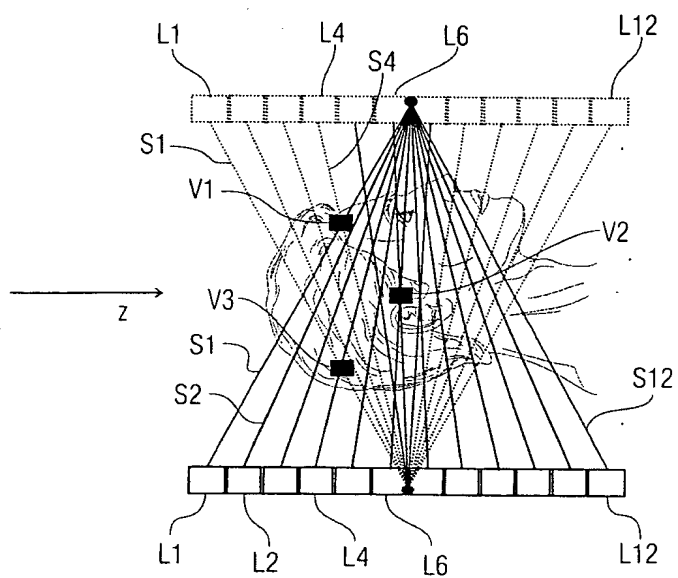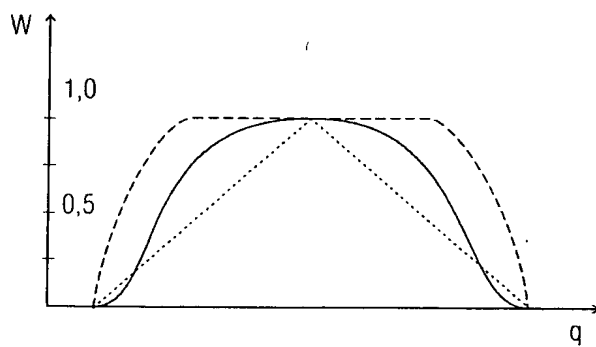
FIG. 4

FIG 1

FIG 2

FIG 3

FIG 4

FIG 5

FIG 6